Inverse dynamics for 3D upper limb movements : a critical evaluation from electromagnetic 6D data obtained in quadriplegic patients.

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A critical evaluation from electromagnetic 6D data obtained in quadriplegic patients.

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***Abstract*— C6 quadriplegic patients lose the ability to use their triceps, but the arm flexors remain intact. However, they are still able to extend the elbow. A previous kinematics study showed that their pointing movements were very similar to those of healthy subjects. We make the hypothesis that the triceps is more likely to be used to ensure elbow stiffness than to provide an active elbow extensor torque. Thus, this study aims at revealing the dynamics of arm pointing movements in quadriplegic patients. We modelled the upper limb from kinematics data acquired from an electromagnetic motion capture device. Inverse dynamics was used to calculate joint torques. Sensitivity analysis was conducted to assess model reliability. We tested the influence of systematic errors in acquisition data and reconstruction errors. Analysis showed mild sensitivity of the joint torque to errors of position or orientation of the sensors and to orientation of the reconstructed segments. However, we could not reliably decompose the total moment at the elbow into the joint coordination system in order to calculate the flexor-extensor moment. Sensitivity analysis proved that inverse dynamics models could be useful in assessing unconstrained 3D upper limb movements. This study emphasises the limits of electromagnetic devices, but suggest ways to improve the recording method, which will have to be validated by further studies evaluating the degree and shape of systematic errors.**

Keywords- Inverse dynamics, motion analysis, upper limb model, sensitivity analysis, spinal cord injury.

I. Introduction

Inverse dynamics computation has been used to discriminate between active torques produced at each joint and interaction torques during non-redundant two-joint motor tasks of the upper limb in a horizontal plane [1][2][3][4]. In these studies, it was demonstrated that the control was hierarchical, with a leading joint (shoulder or elbow) generating the torques responsible for its own movements and producing interaction torques able to move the distal joint (elbow or wrist) and that the role of the active distal torque was limited to fine adjustments to the task details. The use of interaction torques at the shoulder to extend the elbow seems to be linked to a long-lasting skill acquisition of the control of limb dynamics specific of the dominant arm [1][4].

This finding could explain our previous observation that C6 quadriplegic patients with a paralysis of the triceps brachii (but not of elbow flexors or of shoulder muscles) are able to make pointing movements with a smooth trajectory and a joint coordination very similar to that of healthy subjects [5][6]. Our working hypothesis is that the shoulder is also the leading joint for 3D redundant movements, implying that the role of the elbow extensor muscle is more to ensure elbow stiffness than an active torque. Here, we address the first step to test this hypothesis, which is to develop a 3D inverse dynamics model of the upper limb and to test its validity on previous recordings.

II. Methods

A. *Experimental methods*

1) *Subjects*

The data samples were selected from experiments fully described in [6]. The kinematic analysis was performed on data collected in ten subjects: 4 C6 quadriplegic patients with a triceps paralysis (score <2), 2 C7 quadriplegic patients with an incomplete triceps paralysis (score >=3) and 4 healthy subjects. The sensitivity analysis was performed in the data obtained in one healthy subject and one C6 tetraplegic patient.

2) *Experimental set-up and task*

3D aiming movements were performed in several directions by reference to the projection of the centre of the shoulder (Fig.1). The subjects were sitting in the cut-out part of a horizontal table, the height of which was slightly below the level of the elbow, with the trunk immobilised. The target was a red dot (1 cm in diameter) drawn on a 10 cm high support placed about 5 cm out of reach of the arm along one direction indicated by a line drawn on the table. The present analysis was performed on movements directed to the lateral and to an internal target (respectively 0° and 120° by reference to the mediolateral X axis). Aiming was performed with a pointer fixed on the dorsum of the closed hand. The instruction was to aim at a self-paced comfortable speed in the direction of the remote target, as close as possible to it, and to keep this posture for 1-2 s.

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Figure 1. Experimental set-up and task

3) Recording

The 3D movements were recorded with four Fastrak Polhemus sensors (at 30 Hz), attached to the dorsum of the hand (middle part of the third metacarpal bone), to the posterior part of the forearm, to the lateral part of the arm and to the acromion. Each sensor bears a Cartesian coordinate system, the attitude and position of which, relative to a fixed reference frame, is known. The sensors' x-axis was positioned along the longitudinal axis of each segment.

We also measured the length and two diameters of each upper limb segment and the distance of the sensors by reference to bony landmarks.

B. Biomechanical model

1) Kinematic model

Upper limb was modelled as linked chain of three rigid segments (upper arm, forearm and hand). Joints were modelled as pin join with no constraint. Electromagnetic recordings allow the computation of the centre of the glenohumeral joint and of the rotation axes at the elbow and the wrist, based on a preliminary calibration procedure by alternative movements imposed by the experimenter along each DoF [7]. This protocol yields the position and orientation of the rotation axis in seven DoF in the local coordinate system of the sensors fixed on the adjacent segments. To reconstruct the segments, we made the additional assumptions that their upper and lower extremities are defined as the centres of rotation of the proximal and distal joints, respectively, and that segments are linked by their extremities. We were not able to calculate accurately instantaneous centres of rotation due to the slow sampling frequency of Polhemus Fastrak. The shoulder rotation centre was calculated as described in [7], elbow rotation centre was assumed to be on the longitudinal axis of the upper arm, accounting for the distance between shoulder and elbow rotation centres calculated from arm length. The fist is modelled as a sphere. Its extremities were assumed to be on the hand axis (sensor x-axis), accounting for biometrical data. Due to prono-supination, we feared that skin movement artefact might spoil the attitude of the forearm sensor. Thus, we estimated the forearm extremities, over the movement, from the upper arm and hand extremities, knowing the distances from the forearm sensor to these extremities. This resulted in minimization of errors in the forearm sensor's attitude.

Elbow joint coordinate system is attached to the upper arm segment and was calculated as such: X axis, flexion-extension axis calculated as described above; Z axis, along the longitudinal axis of the upper arm; and Y axis to complete the coordinate system.

2) Inverse dynamics model

Body segment parameters (Inertia, mass, centre of mass position) were computed according to [8].

Moments were calculated at the elbow centre of rotation (borne by the upper-arm) by inverse dynamics. Calculi were made thanks to KIHOPSYS Software (Marseille, France) by solving Newton-Euler equation from the hand (We assumed external moment and force were zero at the hand).Total moment was decomposed in the joint coordinated system.

3) Sensitivity analysis

Skin movement artefacts should result in errors in the position of both the elbow centre of rotation and the centre of inertia of the segments. To test the reliability of our model, we evaluated the sensitivity of the total moment at the elbow to such systematic errors in input data. We introduced two sets of errors: 1) bias in segmental inertial parameters, to simulate errors made at the time of reconstruction (the position of the elbow centre of rotation did not change) 2) bias in the coordinates of the sensors, to simulate skin movement artefacts. Segmental errors were either translation of the centre of mass or rotation around the centre of mass. Sensor errors were translation in sensor position and errors in the three dimensions of attitude coordinates. Three increasing distances were used for translations (1, 5 and 10 cm), and three

TABLE I. MAXIMAL DIFFERENCE BETWEEN PERTURBED AND UNPERTURBED TIME PROFILES OF MOMENTS AS A POURCENTAGE OF MEAN TOTAL ELBOW MOMENT FOR LARGEST INTRODUCED ERRORS. TWO SUBJECTS (CONTROL AND QUADRIPLEGIC) AND TWO DIRECTIONS OF MOVEMENT (0° AND 120°) ARE SHOWN.

Maximal error in Moment	Direction 0°		Direction 120°	
	Position Errors	*Orientation Errors*	*Position Errors*	*Orientation Errors*
Control Subject				
Hand sensor	17 %	10 %	10 %	5 %
Forearm sensor	48 %	17 %	23 %	15 %
Upper-arm sensor	65 %	78 %	25 %	90 %
Hand segment	13 %	0 %	15 %	0 %
Forearm segment	54 %	15 %	43 %	10 %
Upper-arm segment	0 %	0 %	0 %	0 %
Quadriplegic Subject				
Hand sensor	41 %	22 %	32 %	18 %
Forearm sensor	57 %	19 %	30 %	32 %
Upper-arm sensor	91 %	95 %	34 %	117 %
Hand segment	22 %	0 %	23 %	0 %
Forearm segment	49 %	15 %	56 %	32 %
Upper-arm segment	0 %	0 %	0 %	0 %

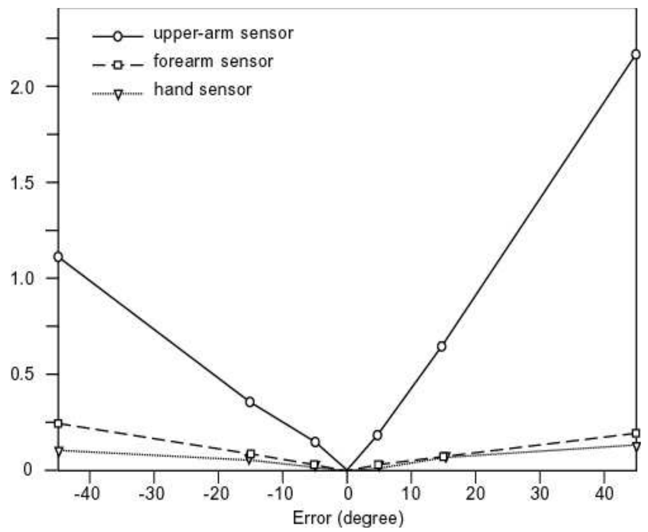

Figure 3. Root mean square error of elbow net moment as a function of introduced errors in attitude of upper arm sensor. *(Control subject, direction 0°)*

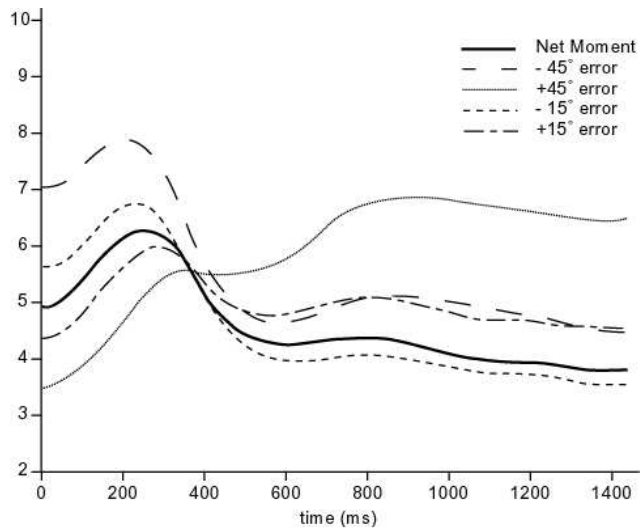

Figure 2. Time profile of elbow total moment.
Variations with introduced errors in attitude of upper arm sensor. *(Control subject, direction 0°)*

increasing angles were used for rotation (5, 15 and 45°). The errors were introduced for each segment or each sensor, while all other segments or sensors were kept unchanged.

Total moment at the elbow centre of rotation was calculated in each condition. Mean total moment was calculated as the mean across the movement. Root mean square error on total moment from unperturbed model was calculated in each condition and compared to mean total elbow moment.

This procedure was repeated for two subjects (a quadriplegic subject and a control subject) and for two directions of the movement in each case (0° and 120°). No pertinent differences were observed between these situations (Table 1). We analysed only the results for the control subject, direction 0°.

III. RESULTS

A. Total moment sensitivity to sensor coordinates errors

The time profile of total moment was qualitatively preserved whatever the perturbed sensor and whether it was attitude or positions errors, and even for high degrees of errors (10 cm or 45°). Total moment profile was only qualitatively perturbed for errors of plus or minus 45° in arm sensor attitude (Fig. 2).

Root mean square errors reveal model sensitivity to each set of errors. The hand sensor has barely any influence on the elbow total moment (0.12 N.m for 5 cm errors, 2.6 % of mean total moment, 0.073 N.m for 15° errors, 1.6 %). The effect of the orientation errors of the sensor fixed on the forearm was greatly reduced by the procedure which calculated forearm extremities (0.084 N.m for 15°, 1.8 %), whereas 5 cm position errors of this sensor result in 0.44 N.m rms errors (9.6 % of mean total moment). As expected, sensitivity to errors were greatest for the sensor fixed on the upper arm (0.61 N.m for 15° attitude errors and 0.56 N.m for 5 cm position errors). However, these errors appear rather mild compared to the mean elbow total moment (13.3 % and 12.2 % of mean total moment, respectively) (Fig. 3).

B. Total moment sensitivity to segmental errors

The time profile of total moment was qualitatively preserved whatever the perturbed segment and whether it was orientation or position errors, even for high degrees of errors (10 cm or 45°)

As expected with the Newton-Euler method, changes in upper arm segment parameters do not have any effect on total moment at the elbow. As the fist is modelled as a sphere, changes in orientation of this segment also have no effect (either supprimé), and errors in position barely perturb the total moment (0.063 N.m for 5 cm errors, 1.4 % of total moment). Biases in orientation and position of forearm segment were of 0.090 N.m for 15° errors, (1.9 % of total moment) 0.34 N.m for 5 cm errors, (7.4 % of total moment), respectively.

C. Projection of the elbow total moment on the flexion-extension axis

In this study, it was not possible to decompose total moment on the three elbow joint axes, because the rotation axes were indeterminate. Indeed, when examining the movement at the elbow, we found that, in addition to the extension movement and a small prono-supination, there was an apparent external rotation of the forearm around the longitudinal axis of the upper arm. In that case the movement could not be possible. As a result, the calculated flexion-extension axis (borne by the upper-arm) does not stay perpendicular to the plane of the real flexion-extension movement. This was verified in the recordings obtained in 10 subjects.

IV. DISCUSSION

The sensitivity analysis showed that the computation of the total moment of the elbow was reliable, since the errors remained relatively low, even for massive perturbations. This suggests that electromagnetic sensors and inverse dynamics models could be used to analyse unconstrained 3D redundant movements of the upper limb.

However, irregularities in the projection of the total elbow moment on the measured flexion-extension axis prevented the computation of the flexor-extensor torque during the movement. This could result either from internal rotation of the upper arm sensor relative to the bone due to skin sliding or from an error in the reconstruction of the forearm axis, since forearm axis and prono-supination axis are not in the same direction. Indeed, a dependence of the measured elbow flexion-extension axis orientation on the forearm pronation or supination has already been observed [7][9]. In the future, the recording methods could be easily improved to overcome this limitation. We suggest that the position of the bony landmarks in the reference frame of the sensors should be measured by a preliminary calibration procedure in order to improve the reconstruction. Another way to improve accuracy could be on the recording side. Sensors on the upper arm could be glue on cast or by carefully chosen sites of attachment close to the bone. In addition, the rotation in the prono-supination axis should be isolated from flexion-extension thanks to a supplementary sensor fixed on the skin along the ulnar crest. As for skin movement artefacts, we do not know the amount of error made by neglecting them, and it is difficult to numerically compensate them, because these are systematic errors. Thus, we want to stress the need for an estimation of the amount and the shape of skin movement artefact. This could achieved by measuring the position of the bony landmarks relative to the sensors through a set of static postures.

According to the present analysis, our existing experimental data cannot be used to test the hypothesis that the triceps brachii acts mainly as an elbow stabilizer in 3D. However, this remains most likely, from observations obtained during 2D movements in healthy subjects [1][2][3][4] as well as in quadriplegic patients [10]. A paralysis of the triceps brachii not only reduces elbow extensor strength but also impedes elbow stiffness. We thus propose that an improvement of limb stability could explain some of the clinical benefits of a musculotendinous transfer [11] and could be further used to develop assistive systems during 3D natural movements [10] [12].

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REFERENCES

[1] Bagesteiro, L. B., & Sainburg, R. L. Handedness: dominant arm advantages in control of limb dynamics. J Neurophysiol, 2002, vol 88, 2408-2421.

[2] Dounskaia NV, Ketcham CJ, Stelmach GE. Influence of biomechanical constraints on horizontal arm movements. Motor Control 2002 vol 6 pp 366-387.

[3] Ketcham, C. J., Dounskaia, N. V., & Stelmach, G. E. (2004). Multijoint movement control: the importance of interactive torques. Prog Brain Res, 143, 207-218.

[4] Sainburg RL, Kalakanis D. Differences in control of limb dynamics during dominant and nondominant arm reaching. J Neurophysiol 2000 vol 83, pp 2661-2675.

[5] Laffont, I., Briand, E., Dizien, O., Combeaud, M., Bussel, B., Revol, M., Roby-Brami, A. Kinematics of prehension and pointing movements in lC6 quadriplegic patients. Spinal Cord, 2000 vol 38, pp 354-362.

[6] Hoffmann, G., Laffont, I., Hanneton, S. Roby-Brami, A., How to extend the elbow with a weak or paralysed triceps? Control of arm kinematics for aiming in C6-C7 quadriplegic patients. Neuroscience, in press.

[7] Biryukova, E. V., Roby-Brami, A., Frolov, A. A., & Mokhtari, M. Kinematics of human arm reconstructed from spatial tracking system recordings. J Biomech, 2000, vol 33, pp 985-995.

[8] Chandler, R. F., Clauser, C. E., McConville, J. T., Reynolds, H. M., Young, J. W. Investigation of inertial properties of the human body. Ohio: AMRL-TR, 1975, 74-137.

[9] Ishizuki, M., Functional anatomy of the elbow joint and threedimensional quantitative motion analysis of the elbow joint. Journal of Japan Orthopaedic Association 1979 vol 53, 989-996.

[10] Perreault EJ, Crago PE, Kirsch RF. Postural arm control following cervical spinal cord injury. IEEE Trans Neural Syst Rehabil Eng 2001; 9: 369-77.

[11] Mulcahey MJ, Lutz C, Kozin SH, Betz RR. Prospective evaluation of biceps to triceps and deltoid to triceps for elbow extension in tetraplegia. J Hand Surg [Am] 2003; 28: 964-71.

[12] Popovic D, Popovic M. Nonanalytical control for assisting reaching in humans with disabilities. In: Winters JM, Crago PE, editors. Biomechanics and neural control of posture and movement. New York: Springer-Verlag; 2000. p. 535-50.